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AUTOMATIC BLOOD PRESSURE MEASURING INSTRUMENT AND METHOD THEREOF

Technical Field

The present invention relates to a blood pressure measuring instrument, and more particularly to an automatic blood pressure measuring instrument and method designed to obtain a pulse wave signal and electrocardiogram (ECG) signals from a pressure sensor and an ECG monitor, to analyze correlation between both signals, to operate a maximum blood pressure and a minimum blood pressure based on the analyzed data, and to output the operated result to a display.

Background Art

As the number of the home health aged has been increased due to the general aging of society, a gradually increasing attention has been drawn to welfare and care for the aged. Thus, the approach of an elderly health care has begun to occur in a new scheme at various angles, and a concentrated attention has been paid to a blood pressure measurement functioning as the basis of health check-up.

Measurement of a blood pressure, as one of current generalized clinical tests, is carried out while a doctor performs an examination or a particular surgical operation. Further, values of blood pressure, which are measured in respective ventricle and atrium of the heart or in a peripheral vascular system, function as a basis of health check-up helpful to enable the doctor to understand an integrated function between vascular and cardiac systems.

A blood pressure refers to one generated when blood flowing a blood vessel acts on a wall of the blood vessel, and is determined by a quantity of the blood and a

resistance of the blood vessel, such as elasticity, expansion, contraction and so on. Measurement of the blood pressure allows for estimation of a function of the heart or the blood vessel. To be more specific, while the heart is contracted, the blood is driven to circulate around the whole body. The pressure of the driven blood, or the maximum blood pressure, represents a contractile force of the heart. The minimum blood pressure when the heart is expanded can be considered as an index indicating how smoothly the blood circulates through the blood vessel.

The human blood vessel consists of an artery through which blood exits from the heart toward the whole body, a vein through which blood enters from the whole body toward the heart, and a capillary vessel interconnecting between the artery and the vein. In general, a pressure in the artery, i.e., an arterial blood pressure is called a "blood pressure." This blood pressure is very different according to size and position of the blood vessel, and is sequentially lowered in the order of an aorta, an artery, an arteriole, a capillary vessel, a veinlet, a vein and a hollow. For this reason, the blood pressure is named for the name of the blood vessel, for example, as an aortic blood pressure, an arterial blood pressure, an arteriole blood pressure and so forth. The arterial blood pressure maintained by heartbeats is one of fundamental, usual clinical symptoms estimating a function between the vascular and cardiac systems, and a factor taking part in perfusion of the entire tissues, or particularly having an important influence on a cerebral blood flow and a coronary blood flow.

As existing blood pressure measuring methods, there are an invasive one and non-invasive one. The invasive method inserts a catheter directly into an artery to measure the blood pressure. However, it requires much trouble and heavy cost in measuring the blood pressure, and has various disadvantages, such as circulatory problem of the blood, infection, blood clot and so on, in the course of inserting the

catheter into the artery. For these reasons, the invasive method has been used in an extremely limited range. By contrast, the non-invasive method mainly makes use of a cuff. However, the non-invasive method is not only very inaccurate, but also is responsible for a tissular trauma. Further, it is impossible to apply to infants or low blood pressure patients, and above all to perform continuous monitoring. Additionally, an electronic hemadynamometer, which has been frequently used at the present time, shows a tendency of its accuracy to be significantly lowered when the blood pressure is less than 70 mmHg.

To measure the blood pressure, there have been many attempts to make use of a pulse wave velocity. Further, there has been devised a blood pressure estimating method using the pulse wave velocity or a pulse arrival time by many persons. However, in the case of using only the pulse wave velocity or the pulse arrival time, it is impossible to ensure accomplishment of reliable blood pressure monitoring

Disclosure of Invention

It is, therefore, an object of the present invention to provide an automatic blood pressure measuring instrument and method, capable of measuring a blood pressure even though infants, low blood pressure patients, intensive care patients, etc. can not be measured by an existing blood pressure measuring instrument, and capable of expanding an application range as a vascular automatic diagnosis device using a pulse wave, including simply blood pressure measurement.

Further, it is another object of the present invention to provide an automatic blood pressure measuring instrument and method, capable not only of continuously measuring systolic and diastolic blood pressures for a short and long time, but also of monitoring for a long time by a non-invasive method.

In order to accomplish these objects, there is provided an automatic blood pressure measuring instrument for measuring and displaying a blood pressure of a subject, comprising: a pressure sensor for obtaining a pulse wave from a wrist of the subject; a pulse wave signal processing section for amplifying, filtering and noise-removing the pulse wave applied from the pressure sensor; an electrocardiogram monitor for measuring a systolic blood pressure and a diastolic blood pressure and converting the measured results into electrical signals; an electrocardiogram signal processing section for amplifying, filtering and noise-removing the converted electrocardiogram measurement signals applied from the electrocardiogram monitor; an A/D converting section for converting the AC signals, which are applied from both the pulse wave signal processing section and the electrocardiogram signal processing section, into DC signals; a controlling section for comparing and analyzing the pulse wave signal and the electrocardiogram signals applied through the A/D converting section to operate the blood pressure of the subject; and a display for displaying the blood pressure of the subject operated at the controlling section.

Further, it is preferred that the automatic blood pressure measuring instrument further comprises a program storing section for storing an operation program of the controlling section, and a data storing section for storing the pulse wave signal and the electrocardiogram signals applied from the A/D converting section for a predetermined time and storing operation data operated at the controlling section.

Here, the pulse wave signal processing section comprises a first impedance matching means for matching impedances of the inputted pulse wave signal and output signal, a pulse wave signal amplifying means for amplifying the signals outputted from the first impedance matching means, and a first notch filter for removing a noise of a commercial frequency from the signals amplified at the pulse wave signal amplifying

means.

Moreover, the first notch filter comprises an OP amplifier for amplifying the signals amplified at the pulse wave signal amplifying means and inputted a non-inverting terminal thereof, a low-pass filter provide on a loop fed from an output terminal of the OP amplifier back to an inverting terminal and for removing the noise of the commercial frequency, a first variable resistor connected in parallel with the non-inverting terminal of the OP amplifier, and a second variable resistor connected in parallel with the low-pass filter, whereby the first notch filter adjusts the cut-off frequency of the applied signals.

Further, the electrocardiogram signal processing section comprises an amplifying section for amplifying the electrocardiogram signals generated from the electrocardiogram monitor, and a filtering section for filtering and noise-removing the signals amplified at the amplifying section.

Meanwhile, the filtering section comprises a fourth low-pass filter for removing a noise from the amplified signals applied from the amplifying section, a third impedance matching means for matching an impedance of the input signal applied from the fourth low-pass filter and an impedance of an output signal, and a second notch filter for removing the noise of the commercial frequency of the signals applied from the third impedance matching means.

The amplifying section comprises a first differential amplifier including a first gain adjusting means for adjusting a gain of the electrocardiogram signals measured from one side of a body of the subject, a second low-pass filter for removing a low band noise from the adjusting signals applied from the first gain adjusting means, and a first electrocardiogram amplifying means from amplifying the signals filtered at the second low-pass filter; a second differential amplifier including a second gain adjusting means

for adjusting a gain of the electrocardiogram signals measured from the other side of a body of the subject, a third low-pass filter for removing a low band noise from the adjusting signals applied from the second gain adjusting means, and a second electrocardiogram amplifying means from amplifying the signals filtered at the third low-pass filter; and a second impedance matching means for matching an impedance with the filtering section when the amplifying signals of the first and second differential amplifiers are applied.

More preferably, the first and second differential amplifiers further comprises an inverse current preventing means connected to an input terminal to which the measurement signals are applied from electrodes of the electrocardiogram monitor.

In order to accomplish these objects, there is provided an automatic blood pressure measuring method for measuring a blood pressure from a wrist of a subject in a non-invasive method using the foregoing construction, comprising the steps of: obtaining, amplifying and filtering a pulse wave from the wrist of the subject; measuring a systolic blood pressure and a diastolic blood pressure, and converting the measured results into electrical signals, and amplifying and filtering the converted results; converting AC signals of the pulse wave and the electrocardiogram into DC signals after the amplifying and filtering steps; comparing the pulse wave and electrocardiogram signals converted at the converting step to operate the blood pressure of the subject; and displaying the blood pressure operated in the operating step.

Here, the comparing and operating step comprising the substeps of: inputting the pulse wave and electrocardiogram signals; comparing the pulse wave and electrocardiogram sensing signals inputted at the measuring step and operating a transition time parameter, an integral parameter, an area parameter and a maximum amplitude parameter; and combining constants representing a change quantity of the

blood pressure according to the transition time parameter, the integral parameter, the area parameter and the maximum amplitude parameter operated at the comparing and operating substep and according to changes of the parameters, and operating the combined results, and operating a maximum blood pressure and a minimum blood pressure.

Further, the transition time parameter is a time interval between maximum amplitudes of a waveform of the pulse wave and waveforms of the electrocardiogram signals. The integral parameter is an integral value of a data value between end points of a selected range of the pulse wave. The area parameter is an integral value of an range joining base lines on both sides of the selected range of the pulse wave. The maximum amplitude parameter is a maximum amplitude within a designated of the integral and area parameters.

In other words, the present invention is characterized in that anyone not a medical expert can measure the blood pressure with ease by sensing the pulse wave at a pressure sensor, sensing and comparing and analyzing systolic and diastolic pressures, measuring maximum and minimum pressures through a series of operation processes, and displaying the measured results on the display.

Brief Description of Drawings

The above objects, features and advantages of the present invention will become more apparent from the following detailed description when taken in conjunction with the accompanying drawings, in which:

Figure 1 is a block diagram illustrating a telephone information offering system on the internet according to an embodiment of the present invention.

FIG. 1 is a perspective view showing an automatic blood pressure measuring

instrument according to a preferred embodiment of the present invention;

FIG. 2 is a bottom view of FIG. 1;

FIG. 3 is a side view of FIG. 1;

FIG. 4 is a block diagram showing an automatic blood pressure measuring instrument of the present invention;

FIG. 5 is a circuit diagram showing a controlling section;

FIG. 6 is a circuit diagram showing a pulse wave signal processing section;

FIG. 7 is a circuit diagram showing an amplifying section of an ECG signal processing section;

FIG. 8 is a circuit diagram showing a filtering section of an ECG signal processing section;

FIG. 9 is a flow chart showing an automatic blood pressure measuring method according to the present invention;

FIG. 10 is a flow chart showing comparing and operating steps;

FIG. 11 is a graph showing each parameter;

FIG. 12 is a graph showing correlation between each parameter according to a change of the systolic blood pressure;

FIG. 13 is a graph showing correlation between each parameter according to a change of the diastolic blood pressure; and

FIG. 14 is a graph tabling accumulative distribution of expectation and observation distribution for a blood pressure algorithm.

Best Mode for Carrying Out the Invention

A preferred embodiment of the present invention will now be described with reference to the accompanying drawings.

FIG. 1 is a perspective view showing an automatic blood pressure measuring instrument according to a preferred embodiment of the present invention. FIG. 2 is a bottom view of FIG. 1. FIG. 3 is a side view of FIG. 1.

A reference numeral 100 indicates an automatic blood pressure measuring instrument, 11 indicates a display, 12 indicates manipulating keys, 13 indicates electrocardiogram (ECG) connection ports, 14 indicates a band, 15 indicates a buckle, 16 indicates a pressure sensor, and 17 indicates an ECG monitor.

The display 11 is for displaying a measured blood pressure. The manipulating keys 12 are for inputting a manipulation signal of a user. The ECG connection ports 13 are for connecting with the ECG monitor 17. The ECG monitor 17 measures a systolic blood pressure and a diastolic blood pressure. The band supports the automatic blood pressure measuring instrument 100. The buckle 15 fixedly positions the automatic blood pressure measuring instrument 100 on a wrist of a subject. The pressure sensor 16 senses pulse waves of the subject.

In order to measure a blood pressure of the subject, first, the pressure sensor 16 is positioned around an artery of the subject. Then, the band 14 is wound around a wrist of the subject, and then the buckle 15 is fastened. Subsequently, the ECG monitor 17 is connected to the ECG connection ports 13. Then, a first electrode (ECG LL, FIG. 7) is fixed to a left foot of the subject, while a second electrode (ECG RA, FIG. 7) is attached to a right foot of the subject.

Therefore, the automatic blood pressure measuring instrument 100 obtains a pulse wave of the subject using the pressure sensor 16. The ECG monitor 17 measures systolic and diastolic blood pressures of the subject. Thus, a controlling section 70 compares and operates both signals to display the maximum and minimum blood pressures on the display 11.

FIG. 4 is a block diagram showing an automatic blood pressure measuring instrument of the present invention.

Of reference numerals, 11 is for a display, 16 is for a pressure sensor, 17 is for an ECG monitor, 20 is for a pulse wave signal processing section, 30 is for an amplifying section, 40 is for a filtering section, 50 is for an ECG signal processing section, 61 is for a program storing section, 62 is for a data storing section, 63 is for an A/D converting section, 70 is for a controlling section, 80 is for an inputting section, 90 is for an interface, and 200 is for a computer.

The pulse wave signal processing section 20 amplifies and filters a pulse wave applied from the pressure sensor 16. The amplifying section 30 amplifies a signal applied from the ECG monitor 17. The filtering section 40 filters an applied low band signal. The program storing section 61 is stored with a drive program together with set data. The data storing section 62 is stored with a measured signal together with operated data. The A/D converting section converts an AC signal into a DC signal. The controlling section operates the maximum blood pressure and the minimum blood pressure. The inputting section 80 is inputted by a manipulation signal of a user. The interface 90 is connected with an external instrument.

The pressure sensor 16 fixed to the wrist of the subject generates a pulse wave caused by a pressure of blood flowing through the artery. That is, the pressure sensor 16 fixed over the blood vessel is subjected to impetus which the blood lends to the blood vessel, so that the pressure sensor 16 generates the pulse wave. The pulse wave applied from the pressure sensor 16 is inputted into the pulse wave signal processing section 20, and is amplified there. As a result, the pulse wave is subjected to filtering of a low band signal and removal of a noise. Then, the signal from which the noise is removed is applied to the A/D converting section 63, and is subjected to conversion

from an AC signal to a DC signal, and then is applied to the controlling section 70.

Further, the ECG monitor 17 measures a systolic blood pressure as well as a diastolic blood pressure to generate ECG signals. The generated ECG signals are applied to and amplified by the ECG signal processing section 50, and are applied to the A/D converting section 63. The A/D converting section 63 converts the applied AC signals into DC signals to apply to the controlling section 70.

Here, the controlling section 70 is capable of performing arithmetical operation of data of 8 bits as shown in FIG. 5. To this end, an 8051 microprocessor having a 16-bit data address is preferably used. The controlling section 70 has four input/output ports, so that it is capable of directly receiving and outputting data of the data storing section 62 and the A/D converting section 63 relative to the outside. Further, the controlling section 70 has a built-in serial port, so that it is capable of directly receiving and outputting data from/to the computer 200 through the interface 90 using the serial port. The controlling section is capable of not only storing a program to the program storing section 61, but also performing various records related to bit operation, controlling, etc. using an SFR (special function register).

Here, it is preferred that the data storing section 62 makes use of an SRAM in order to avoid a work such as a refresh. The data storing section 62 stores data obtained by the A/D converting section 63 for a predetermined time under the control of the controlling section 70, and performs comparison and analysis of the pulse wave and the ECG signals, which are applied based on the drive program stored at the program storing section 61. Then, the data storing section 62 operates a transition time parameter (or ΔT parameter) c, an integral parameter a, an area parameter b and the maximum amplitude parameter (or Max) d, and then applies the operated parameters a, b, c and d to a blood pressure algorithm, which will be mentioned below, and finally

operates values of the maximum and minimum blood pressures. The controlling section 70 stores the operated data to the data storing section 62, and controls the display 11 to force the operated values of the maximum and minimum blood pressures to be displayed. Additionally, the controlling section 70 is capable of operating a pulse rate or frequency of the subject based on the applied measurement signals. This operation algorithm for the pulse rate or frequency is preferably stored at the program storing section 61.

Here, the controlling section 70 checks whether there is connection to an external instrument by means of the interface 90. If the connection to an external instrument such as the computer 200, etc. is present, the controlling section 70 transmits the operated data to the computer 200 through the interface 90.

A construction of the pulse wave signal processing section 20 and the ECG signal processing section 50 as mentioned above is shown in detail in FIGs. 6 to 8. Hereinafter, the construction will be described in detail with reference to FIGs. 6 to 8.

FIG. 6 is a circuit diagram of the pulse wave signal processing section of FIG. 4.

Of reference numerals, 21 indicates a first impedance matching means, 22 indicates a pulse wave signal amplifying means, and 23 indicates a first notch filter.

The first impedance matching means 21 matches an impedance of the pulse wave applied from the pressure sensor 16 to an impedance of an output terminal. The pulse wave signal amplifying means filters a low band signal to remove a noise. The first notch filter 23 removes a noise of a commercial frequency.

When the measurement signals are applied from the pressure sensor 16, these signals are applied to inverting and non-inverting terminals of a first OP amplifier (OP1) through first and second resistors (R1 and R2) connected in parallelism, and then their voltages are amplified by the first OP amplifier OP1. Here, because the applied

measurement signals of the pressure sensor 16 has a higher impedance at an output terminal of the first OP amplifier OP1, both the first resistor R1 and the first OP amplifier OP1 match an impedance at each input terminal to that at each output terminal. That is, according to a rule of voltage distribution, when a resistance of the R1 is increased, an input voltage is lowered. The lowered input voltage is amplified at the first OP amplifier OP1, so that the impedance of the R1 is matched with that of a circuit connected to the first OP amplifier OP1. Therefore, the signals outputted from the first OP amplifier OP1 through the process as mentioned above are subjected to the foregoing impedance matching process and the amplifying process through a fourth resistor R4, and a second OP amplifier OP2.

The signals outputted from the first impedance matching means 21 pass through a sixth resistor R6, of a first low-pass filter 22, and are subjected to removal of a low band signal between 20 and 40 Hz by means of a seventh resistor R7, and a second capacitor C2, and are inputted into an inverting terminal of a third OP amplifier OP3. Then, the signals inputted into the third OP amplifier OP3 are subjected to amplification, and pass through a tenth resistor R10, and are filtered by an eleventh resistor R11 and a fourth capacitor C4. Subsequently, the secondary filtered pulse waves are applied to a fourth OP amplifier OP4. The OP4 amplifies the imputed signals to apply to the first notch filter 23.

The first notch filter 23 includes a fourteenth resistor R14 connected to an output terminal of the fourth OP amplifier OP4 in series, a first variable resistor, VR1 connected to the R14 in parallel, a fifth OP amplifier OP5 having an inverting terminal connected to the R14 in series and an output terminal designed to perform negative feedback to a non-inverting terminal, a fifth capacitor C5 connected to a negative feedback loop of the fifth OP amplifier OP5 in series, a sixth capacitor C6 connected to

the C5 in parallel and to the non-inverting terminal of the fifth OP amplifier OP5 in series, a fifteenth resistor R15 connected to the C5 in parallel, and a second variable resistor, VR2 having one side grounded and the other side connected to a sixteenth resistor R16, which is connected to the R15 in parallel.

The pulse wave signal applied from the first low-pass filter 22 is inputted into the first notch filter 23, particularly, the non-inverting terminal of the fifth OP amplifier OP5 via the R14 and the VR1. The pulse wave signal inputted into the fifth OP amplifier OP5 is subjected to amplification and are outputted. At this time, the R15 and the C5, which are connected in parallel on the negative feedback loop, remove a noise at a commercial frequency of 60 Hz, which is subjected to feedback at the output terminal of the fifth OP amplifier OP5. Here, the VR1 and the VR2 are each varied to adjust the commercial frequency to 60 Hz, so that the commercial frequency can be matched to a commercial frequency of a system connected with the output terminal of the first notch filter 23.

FIG. 7 is a circuit diagram showing an amplifying section of an ECG signal processing section.

Of reference numerals or symbols, 30a is for a first differential amplifier, 30b is for a second differential amplifier, an ECG LL is for a first electrode, an ECG RA is for a second electrode, 31 is for a first inverse current preventing means, 33 is for a first ECG signal amplifying means, 34 is for a second low-pass filter, 35 is for a second inverse current preventing means, 36 is for a second gain adjusting means, 37 is for a second ECG signal amplifying means, 38 is for a third low-pass filter, and 39 is for a second impedance matching means.

The first and second inverse current preventing means 31 and 35 prevent an inverse current generated at an input power source. The first and second ECG signal

amplifying means 33 and 37 amplify inputted signals. The second and third low-pass filters 34 and 38 filter signals of a low band frequency among inputted signals. The second impedance matching means 39 matches an impedance of inputted signals to that of outputted signals.

In the first differential amplifier 30a, the first gain adjusting means 32 is connected with the first electrode, ECG LL of the ECG monitor 17 in series. The inverse current preventing means 31 is connected between the ECG LL and the first gain adjusting means 32, and includes first and second power source terminals, +BS1 and –BS1 connected in parallel, and first and second diodes, D1 and D2 connected to the first and second power source terminals, +BS1 and –BS1 in an opposite direction to each other. Here, the first gain adjusting means 32 includes a sixth OP amplifier OP6. A third power source, +BM is applied to a terminal 1 of the sixth OP amplifier OP6, and a fourth power source, -BM is applied to a terminal 2 of the sixth OP amplifier OP6. The third power source, +BM is connected in parallel with eighth and ninth capacitors C8 and C9, each of which is grounded on one side. The fourth power source, -BM is connected in parallel with tenth and eleventh capacitors C10 and C11, each of which is grounded on one side. Further, the first gain adjusting means 32 is connected in parallel with the second impedance matching means 39, the first ECG signal amplifying means 33 and the second low-pass filter 34. Here, the first ECG signal amplifying means 33 includes a seventh OP amplifier OP7, an output terminal of which is subjected to negative feedback to an inverting terminal thereof. A negative feedback loop of the inverting terminal of the seventh OP amplifier OP7 is connected with an output terminal of the second low-pass filter 34. Further, the seventh OP amplifier OP7 of the first ECG signal amplifying means 33 has the output terminal connected to an output terminal of the first gain adjusting means 32.

In the second differential amplifier 30b, the second gain adjusting means 36 is connected with the second electrode, ECG RA of the ECG monitor 17 in series. The second inverse current preventing means 35 is connected between the ECG RA and the second gain adjusting means 36, and includes fifth and sixth power source terminals, +BS2 and –BS2 connected in parallel, and third and fourth diodes, D3 and D4 connected to the fifth and sixth power source terminals, +BS2 and –BS2 in an opposite direction to each other. Here, the second gain adjusting means 36 includes a ninth OP amplifier OP9. A seventh power source, -BM2 is applied to a terminal 1 of the ninth OP amplifier OP9, and an eighth power source, +BM2 is applied to a terminal 4 of the ninth OP amplifier OP9. The seventh power source, -BM2 is connected in parallel with sixteenth and seventeenth capacitors C16 and C17, each of which is grounded on one side. The eighth power source, +BM2 is connected in parallel with fourteenth and fifteenth capacitors C14 and C15, each of which is grounded on one side. Further, the second gain adjusting means 36 is connected in parallel with the second impedance matching means 39, the second ECG signal amplifying means 37 and the third low-pass filter 38. Here, the second ECG signal amplifying means 37 includes a tenth OP amplifier OP10, an output terminal of which is subjected to negative feedback to an inverting terminal thereof. A negative feedback loop of the inverting terminal of the first OP amplifier OP10 is connected with an output terminal of the third low-pass filter 38. Further, the ninth OP amplifier OP9 of the second ECG signal amplifying means 37 has the output terminal connected to an output terminal of the second gain adjusting means 36.

The first and second differential amplifiers 30a and 30b constructed as the foregoing are connected in parallel not only by seventh and thirteenth capacitors C7 and C13 at the rear ends of sixteenth and twenty-seventh resistors R16 and R27 of the input

line, but also by seventeenth and twenty-eighth resistors R17 and R28 on the negative feedback loops, each of which is connected to each inverting terminal of the sixth OP amplifier OP6 and OP9. Further, the first differential amplifier 30a is connected to an inverting terminal of the second matching means 39, while the second differential amplifier 30b is connected to a non-inverting terminal of the second matching means 39.

Signals applied at the first electrode, ECG LL are applied to the first gain adjusting means 32 through the sixteenth resistor R16. Here, since ECG signals applied at the first electrode, ECG LL typically have high impedances, their impedances are not matched with system-side ones, so that there is possibility to give a fatal damage to the system. Thus, input impedances are matched with the system side by the first gain adjusting means 32, and thereby a gain of output signals is adjusted to be lowered. Here, it is preferred that the R16 has a high resistance value in order to apply a rule of voltage distribution. Therefore, the measurement signals applied from the first electrode, ECG LL are subjected to voltage division, so that a gain of the signals outputted from the sixth OP amplifier OP6 is lowered. Further, a current is applied to the first and second power source terminals, +BS1 and –BS1, so that the measurement signals have an increased current. The power source terminals are provided with the first and second diodes, D1 and D2 respectively, and thus the circuit is prevented from being damaged by an inverse current.

The measurement signals outputted from the first gain adjusting means 32 are inputted into a non-inverting terminal of the first ECG signal amplifying means 33 and a non-inverting terminal of the second low-pass filter 34. Here, the second low-pass filter 34 has a feedback loop from an output terminal of the eighth OP amplifier OP8 through the twelfth capacitor C12 to a terminal 1 of the eighth OP amplifier OP8. At this time, signals ranging from 20 to 40 Hz are filtered by the twelfth capacitor C12 on

the feedback loop.

Therefore, the measurement signals filtered at the second low-pass filter 34 are inputted into an inverting terminal of the seventh OP amplifier OP7 of the first ECG signal amplifying means 33, and then are subjected to negative feedback from an output terminal of the seventh OP amplifier OP7 through a twenty-third resistor R23 to the inverting terminal of the seventh OP amplifier OP7, so that they are amplified and applied to the inverting terminal of the second matching means 39.

Signals applied at the second electrode, ECG RA are applied to the second gain adjusting means 36 through the twenty-seventh resistor R27. Here, the second electrode, ECG RA is considered as an external resistor as mentioned above. Since the second electrode, ECG RA and the twenty-seventh resistor R27 are connected in parallel, a circuit line tapped between them is connected to a non-inverting terminal of the ninth OP amplifier OP9. Therefore, according to voltage distribution of the second electrode, ECG RA and the twenty-seventh resistor R27, the measurement signals applied from the second electrode, ECG RA are subjected to voltage division and are inputted into the ninth OP amplifier OP9, so that a gain of the signals is lowered. At this time, a current is applied to the fifth and sixth power source terminals, +BS2 and –BS2, so that the measurement signals have an increased current. The power source terminals, +BS2 and –BS2 are each provided with the second inverse current preventing means 35 consisting of the third and fourth diodes, D3 and D4, and thus the circuit is prevented from being damaged by an inverse current.

The measurement signals outputted from the second gain adjusting means 36 are inputted into a non-inverting terminal of the second ECG signal amplifying means 37 and a non-inverting terminal of the third low-pass filter 38. Here, the third low-pass filter 38 has a feedback loop from an output terminal of the eleventh OP amplifier

OP11 through the eighteenth capacitor C18 to a terminal 1 of the eleventh OP amplifier OP11, and thereby signals between 20 and 40 Hz are filtered.

Therefore, the measurement signals filtered at the third low-pass filter 38 are inputted into an inverting terminal of the tenth OP amplifier OP10 of the second ECG signal amplifying means 37, and then are subjected to negative feedback from an output terminal of the tenth OP amplifier OP10 through a thirty-fourth resistor R34 to the inverting terminal of the tenth OP amplifier OP10, so that they are amplified and applied to the non-inverting terminal of the second matching means 39.

Thus, an output terminal of the twelfth OP amplifier OP12 of the second matching means 39 is connected with a fortieth resistor R40 and forms a negative feedback loop together with an inverting terminal of the twelfth OP amplifier OP12. A voltage inputted into a thirty-seventh resistor R37 provided on an input side of the twelfth OP amplifier OP12 and a voltage inputted into the fortieth resistor R40 of the negative feedback loop are divided and inputted into the twelfth OP amplifier OP12, thus being matched with an impedance of the output terminal and applied to a filtering section 40.

FIG. 8 is a circuit diagram showing the filtering section 40 of the ECG signal processing section 50.

Of reference numerals, 40 indicates the filtering section, 41 indicates a fourth low-pass filter, 42 indicates a third impedance matching means, and 43 is a second notch filter.

The fourth low-pass filter 41 removes a noise of low band from applied signals. The third impedance matching means 42 matches an impedance of an input terminal with that of an output terminal. The second notch filter 43 removes a noise of a commercial frequency.

In the fourth low-pass filter 41, an input terminal is connected in parallel with a nineteenth capacitor C19, and is connected in series with a forty-first resistor R41, a forty-third resistor R43 and an inverting terminal of a thirteenth OP amplifier OP13. A twentieth capacitor C20 is connected in parallel between the forty-first resistor R41, the forty-third resistor R43. A forty-fourth resistor R44 is connected to the nineteenth and twentieth capacitors C19 and C20 on one side and to a non-inverting terminal of the thirteenth OP amplifier OP13 on the other side. Here, the thirteenth OP amplifier OP13 has a negative feedback loop from an output terminal to the non-inverting terminal. Here, a forty-second resistor R42 and a twenty-first capacitor C21 are each connected in parallel to the negative feedback loop.

The third impedance matching means 42 has a forty-fifth resistor 45 R45 connected in series with the output terminal of the fourth low-pass filter 41. The forty-fifth resistor 45 R45 is connected in series to an inverting terminal of a fourteenth OP amplifier OP14. A forty-sixth resistor R46 is connected to a non-inverting terminal of the fourteenth OP amplifier OP14 on one side, and is grounded on the other side. The fourteenth OP amplifier OP14 has a terminal 1 to which a commercial power source, VCC1 is applied. The commercial power source, VCC1 is connected in parallel with twenty-fourth and twenty-fifth capacitors C24 and C25. The fourteenth OP amplifier OP14 has a terminal 4 to which another commercial power source, VCC2 is applied. The commercial power source, VCC2 is connected in parallel with twenty-second and twenty-third capacitors C22 and C23. Thus, a noise is removed from the applied power source. The fourteenth OP amplifier OP14 has a negative feedback loop from an output terminal to an inverting terminal. A forty-seventh resistor 47 is connected to the negative feedback loop of the fourteenth OP amplifier OP14.

The second notch filter 43 has a forty-ninth resistor R49, one side of which is

connected in series with the third impedance matching means 42 and the other side is connected in series with a non-inverting terminal of a fifteenth OP amplifier OP15. Here, a fourth variable resistor, VR4 is connected in parallel between the forty-ninth resistor R49 and the non-inverting terminal of the fifteenth OP amplifier OP15. Further, the fifteenth OP amplifier OP15 is formed with a feedback loop from an output terminal to an inverting terminal. A fiftieth resistor R50 and a twenty-seventh capacitor C27 are each connected in parallel to the feedback loop of the fifteenth OP amplifier OP15. A third variable resistor, VR3 is connected in parallel with the fiftieth resistor R50 through a forty-eighth resistor R48. A twenty-sixth capacitor C26 is connected in parallel with the fiftieth resistor R50 and the twenty-seventh capacitor C27 on one side, and with the inverting terminal of the fifteenth OP amplifier OP15 on the other side.

From the measurement signals applied at the amplifying section 30, low band signals between 20 and 40 Hz are filtered by the forty-first and forty-third resistors R41 and R43, and by the nineteenth and twentieth capacitors C19 and C20. The filtered signals are applied to and amplified by the thirteenth OP amplifier OP13. Here, the amplified signals are applied to the negative feedback loop of the output terminal again, and are filtered by the forty-second resistor R42 and the twenty-first capacitor C21, and are applied to the inverting terminal of the thirteenth OP amplifier OP13.

As mentioned above, the signals filtered at the fourth low-pass filter 41 are inputted into the inverting terminal of the fourteenth OP amplifier OP14 through the forty-fifth resistor R45 of the third impedance matching means 42, so that a voltage amplified to the output terminal is inputted into the inverting terminal again along the negative feedback loop. Therefore, according to a rule of voltage distribution, the inputted signals are subjected to voltage division at the forty-fifth and forty-seventh

resistors R45 and R47. The voltage-divided signals are amplified again and outputted, thus being matched with an impedance of the second notch filter 43.

The signals outputted from the third impedance matching means 42 are applied to the non-inverting terminal of the fifteenth OP amplifier OP15 through the forty-ninth resistor R49 of the second notch filter 43. Here, the fourth variable resistor VR4 adjusts a commercial frequency of the inputted signals. That is, the commercial frequency is 60 Hz, but from 58 to 59 Hz in reality, so that it is adjusted and matched at the fourth variable resistor VR4. The signals amplified at the fifteenth OP amplifier OP15 are subjected to negative feedback, and are filtered at the fiftieth resistor R50 and the twenty-seventh capacitor C27, and are subjected to removal of a noise of the commercial frequency. At this time, the signals are adjusted to the commercial frequency by adjustment of the third and fourth variable resistors VR3 and VR4.

FIG. 9 is a flow chart showing an automatic blood pressure measuring method according to the present invention. An operation of the foregoing construction of the present invention will be described in detail with reference to FIG. 9.

The automatic blood pressure measuring instrument 100 according to the present invention is fixed to the right wrist of the subject so as to position the pressure sensor 16 over the artery of the subject. The ECG monitor 17 is connected to the ECG connection ports 13 provided on one side of the automatic blood pressure measuring instrument 100. The first electrode ECG LL of the ECG monitor 17 is fixed to the left ankle of the subject, while the second electrode ECG RA is fixed to the right arm.

A power switch provided on the upper surface of the automatic blood pressure measuring instrument is turned on (S11), the display 11 is operated (S12), and the pressure sensor 16 and the ECG monitor 17 generate sensing signals (S13 and S14). Therefore, the pulse wave signal sensed at the pressure sensor 16 is applied to the pulse

wave signal processing section 20, so that the signals outputted from the pressure sensor 16 have an impedance matched with that of the signals outputted from the first impedance matching means 21, and are amplified by the pulse wave amplifying means 22, and are applied to the first notch filter 23 again, and are subjected to noise removal at the commercial frequency of 60 Hz (S15). Then, the AC signals free from a noise are converted into DC ones at the A/D converting section 63 and then applied to the controlling section 70 (S16).

Further, the ECG measurement signals generated at the ECG monitor 17 are applied to and amplified at the amplifying section 30. That is, the signals measured at the left leg are amplified at the first differential amplifier 30a, and are applied to the inverting terminal of the second impedance matching means 39, while the signals measured at the right arm are amplified at the second differential amplifier 30b, and are applied to the non-inverting terminal of the second impedance matching means 39, so that both signals are matched with an impedance of the filtering section 40 and are applied to the filtering section 40.

Thus, the amplified signals are applied to the filtering section 40, the fourth low-pass filter 41 allows for pass only some of the amplified signals belonging to a predetermined band, but removes the rest. The filtered ECG measurement signals are applied to the third impedance matching means 42. The third impedance matching means 42 performs buffering in order to match the inputted signals to the second notch filter 43 on the output side of the filtering section 40. The buffered signals are applied to the second notch filter 43, and subjected to noise removal of the commercial frequency of 60 Hz, and applied to the A/D converting section 63, so that they are subjected to conversion from the AC signals to DC signals and applied to the controlling section (S15 and S16).

The controlling section 70 stores the applied measurement signals on the data storing section 62 for a predetermined time, and reads out them, and compares and operates respective data, and obtains a transition time (ΔT) parameter, an integral parameter and an area parameter. The controlling section 70 controls to apply an operation algorithm stored at the program storing section 61 to each parameter, to operate the maximum and minimum blood pressures, and to display these blood pressures on the display 11 (S17 and S18).

Further, the controlling section 70 operates pulse number and diagnosis result using the measurement signals stored at the data storing section 62 for a predetermined time. That is, the controlling section 70 controls to apply the measurement signals to an operation algorithm for the pulse number and diagnosis result stored at the data storing section 62, to operate the pulse rate and frequency to display them on the display 11.

Here, the blood pressure, the pulse number and the pulse diagnosis result are displayed through the LCD (Liquid Crystal Display) 11, so that limitation to a displayed quantity is removed. The display 11 is adapted to display the maximum and minimum blood pressure measured by the automatic blood pressure measuring instrument, as well as the pulse number and the present pulse diagnosis result. Further, the display 11 is designed to display cardiovascular disease codes, an output mode of which is developed based on expected cardiovascular diseases. Table 1 shows items displayed on the display 11 including the disease codes as one example.

Table 1: example of expected disease codes

symptom	**code**	**symptom**	**code**
normal	Nomal	unstable high blood pressure	H-Case S
high normal blood	H-Case 0	first period of high blood	H-Case 1

pressure		pressure	
low blood pressure	L-Case	second period of high blood pressure	H-Case 2
slow pulse	B-Case	third period of high blood pressure	H-Case 3
weak pulse	T-Case	fourth period of high blood pressure	H-Case 4
weak beat of artrim	A-Case	re-measuring request	ERROR

FIG. 10 is a flow chart showing comparing and operating steps. FIG. 11 is a graph showing each parameter. The comparing and operating steps will be described in detail with reference to FIGs. 10 and 11.

When the measurement signals generated from the pressure sensor 16 and the ECG monitor 17 are temporarily stored on the data storing section 62 through the A/D converting section 63 (S21 and S22), the controlling section 70 reads out the measurement signals stored on the data storing section 62 after a predetermined time lapses away, and compares waveforms of the two signals and operates each parameter. That is, as shown in FIG. 11, the pulse wave signal and the ECG measurement signal are compared and analyzed, and each parameter is operated. Of reference symbols, a is the integral parameter, b is the area parameter, c is the transition time (ΔT) parameter, and d is the maximum amplitude parameter.

Thus, the controlling section 70 reads out the data stored temporarily on the data storing section 62, and selects a predetermined zone, and operates the integral parameter a according to the following equation with an integral value for data value between end points of the selected zone (S24).

$$< PSTYLELSPACE = 130 > f_{output}(n) = \sum_{k=1}^{n-1} f_{input}(k) + [(f_{input}(n-1) + f_{input}(n))/2] * \Delta T$$

Further, the controlling section 70 sets a base line joining end points of the area selected within the predetermined zone, and integrates a zone of the upper side of the base line to obtain the area parameter b (S25). This calculation value will be always a

positive value, and can be measured as the whole area between the waveform and the base line, and can represent the area in a unit of the amplitude and the horizon, and can be calculated using a formula given below.

$$< PSTYLELSPACE = 130 > f_{output}(n) = \sum_{k=1}^{n-1} \left(\left| f_{input}(n-1) - y(n-1) \right| + \left| (f_{input}(n-1) + y(n))/2 \right| \right) * \Delta T$$

Here, f() represents a value of data, and y() is a value of the line joining end points, and ΔT is a sampling interval of the horizon.

The controlling section 70 operates a time interval between the maximum amplitudes of an ECG waveform and a waveform detected from the pressure sensor fixed to the wrist, and operates the ΔT parameter c (S26), and operates the maximum amplitude within a designated range of the integral and area parameters a and b, and thus operating the maximum amplitude parameter d (S27). The maximum amplitude parameter d derives a change of the maximum amplitude value of the detected waveform according to a pressure change.

Here, the respective parameters a, b, c and d operated in the foregoing way represent constant tendencies of the systolic and diastolic blood pressures. This linear change is shown in FIGs. 12 and 13.

FIG. 12a is a graph showing a value of the transition time parameter c according to a change of the systolic blood pressure. It can be seen that as the blood pressure becomes higher, a time interval between maximum amplitudes becomes narrower. FIG. 12b is a graph showing a change of the integral parameter a. It can be seen that as the blood pressure becomes higher, an integral value is decreased within a predetermined range. Thus, a value of the integral parameter a is also decreased. Further, FIG. 12c is a graph showing a change of the area parameter b. As the blood pressure becomes higher, a value of data is increased within a predetermined area. Thus, a value of the

area parameter b is increased. FIG. 12d is a graph showing a change of the maximum amplitude parameter d. As the blood pressure becomes higher, a value of the maximum amplitude parameter d is also increased.

Further, FIG. 13 is a graph showing a change of each parameter according to a change of the systolic blood pressure. FIG. 13a is a graph showing a value of the transition time parameter c, which is linearly decreased as the blood pressure becomes higher. FIG. 13b is a graph showing a change of the value of the area parameter b, in which the value of the area parameter b is decreased as the blood pressure becomes higher.

As for correlation of each parameter shown in FIGs. 12 and 13, as the systolic blood pressure is increased, the integral and transition time parameters a and c result in a linear decrease, and the area and maximum amplitude parameters b and d result in a linear increase. In the diastolic blood pressure, the transition time and area parameters c and b result in a linear decrease according to the blood pressure.

Using the correlation between the blood pressure and the respective parameters a, b, c and d, the present invention proposes an operation algorithm as follows:

Maximum blood pressure (systolic blood pressure)

$P= 919.121Ar+17.157Max-98.26Int + 161.736D\_Dt$

Minimum blood pressure (diastolic blood pressure)

$P= 146.161-78.903\ D\_Dt-442.904\ D\_Ar$

where P : pressure (mmHg), Dt and D_Dt : ΔT (sec), Ar and D_Ar : area, and Int : Integral.

The maximum blood pressure refers to the systolic blood pressure. The formula for the maximum blood pressure is derived using the correlation between each parameter and the blood pressure. The minimum blood pressure refers to the diastolic

blood pressure. The formula for the minimum blood pressure is derived using relationship between the blood pressures according to changes of the transition time and area parameters c and b.

Further, in the algorithm, a series of constants by which the respective parameters a, b, c and d are multiplied refer to quantities by which the parameters a, b, c and d are varied according to the blood pressure, and represent a specific weight of each parameter according to the change of the blood pressure. That is, the constant is for statistically representing a change rate of the blood pressure as the parameters a, b, c and d are changed.

Therefore, the controlling section 70 controls to apply the respective parameters a, b, c and d operated in the foregoing step to the operation algorithm, and to operate the systolic blood pressure and the diastolic blood pressure, and to display the operated result on the display 11 (S28 and S29).

FIG. 14 is a graph tabling accumulative distribution of expectation and observation distribution for a blood pressure algorithm using the forgoing each parameter.

As shown, the graph takes a straight line when two values are same. By observing how standardized residuals are distributed relative to an expected straight line, two distribution can be compared. The result is that, because the standardized residuals are near normal distribution, the blood pressure measurement using the automatic blood pressure measuring instrument and method according to the present invention has a very high prediction degree.

The automatic blood pressure measuring instrument of the present invention using a pressure sensor such as a piezo sensor measures a blood pressure using correlation between a pulse wave and an ECG, and can be used to measure blood

pressures of all persons including infants, low blood pressure patients, intensive care patients, because it makes use of various parameters between the pulse wave and the ECG, such as a transition time parameter, an integral parameter, an area parameter and a maximum amplitude parameter.

Additionally, it is possible to measure a blood pressure without inserting a catheter into an artery. Thus, it is possible to prevent various disadvantages which has attracted attention in an existing invasive blood measuring method, such as a circulatory problem, infection, blood clot and so on, and to measure an accurate blood pressure with more ease and safety. Further, it is possible to change a concept on an existing uncomfortable and inaccurate blood pressure measurement. It is expected that a future-oriented and high-tech and multifunctional blood pressure measuring instrument will be distributed at an exponential speed, and that request for industrialization will be rapidly increased, and that due to the advent of brand-new multifunctional microwave hemadynamometer, technical application to a domestic medical instrument field will be much expanded.

While the invention has been shown and described with reference to certain preferred embodiments thereof, it will be understood by those skilled in the art that various changes in form and details may be made therein without departing from the spirit and scope of the invention as defined by the appended claims.